THE EFFECTS OF VARIABLE HELMET WEIGHT AND SUBJECT BRACING ON NECK LOADING DURING FRONTAL -GX IMPACT

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ABSTRACT

Helmet-mounted systems (HMS), such as night vision goggles and helmet-mounted displays, are designed to enhance pilot performance. Using HMS, however, may also affect pilot safety by increasing the potential for neck injury during ejection due to the increase in dynamic forces generated in the cervical spine as a result of the change in helmet inertial properties. Pilot bracing techniques may also have an effect on ejection injury risk by allowing some of the neck forces to be off-loaded during the acceleration impact phases. A series of tests were conducted on the AFRL/HEPA Horizontal Impulse Accelerator (HIA) using human subjects to investigate the effects of helmet inertial properties and bracing techniques on human response to short-duration frontal impacts of variable magnitude. Head accelerations were measured and neck loads and moments were calculated to compare the head and neck responses using helmets of varying weight. Headrest loads were recorded to monitor and evaluate subject bracing. The neck loads and helmet weights were also extrapolated to higher levels in order to estimate injury thresholds for pilots wearing even heavier helmets at maximum seat accelerations. The results of this study will be used to establish head/neck injury criteria for helmet-mounted systems and to improve bracing techniques to minimize pilot injury during ejections.

BACKGROUND

Tests by Perry at the Air Force Research Lab's Biomechanics Branch (AFRL/HEPA) from 1991 through 1997 have evaluated the effects of variable helmet inertial properties on the biodynamic response of male and female human volunteers exposed to vertical (+Gz) accelerations using the Vertical Deceleration Tower (VDT).[1, 4-6, 8] A recent study by Perry and Buhrman investigated the effects of varied helmet weight on human response during lateral +Gy Impact on the Horizontal Impulse Accelerator.[7] Another recent study by Pint explored the effects of varied helmet weight on human neck response during retraction using the Body Positioning and Restraint Device (BPRD).[9] The objective of this study was to provide additional human dynamic response data from a frontal (-Gx) impact environment with a variable weight helmet. This is required to complete the development of multi-axial cervical injury criteria for the three coordinate axes, and to continue the development of head/neck biodynamic models. In particular, the results of this program will contribute to the development of design guidelines for the safe use of helmet-weighted systems and provide information on optimal pilot bracing techniques.

METHODS

A series of short-duration frontal impact tests were completed at Wright Patterson AFB using the Horizontal Impulse Accelerator (HIA).

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Thirty-four subjects, sixteen females and eighteen males, were tested with approval obtained from the Wright Site Institutional Review Board. The HIA has a 240-foot long track and a 24-inch diameter pneumatic actuator and operates on the principle of differential gas pressures acting on both surfaces of a thrust piston in a closed cylinder. The acceleration profile approximated a half-sine pulse with rise-time and pulse duration of 75 and 150 ms, respectively. Peak sled acceleration levels of 6, 7, 8 and 10 G were generated with total head supported weight ranging from 0 (no helmet) to 4.5 lbs. The acceleration levels had been previously tested, and were well tolerated by the volunteer human subjects.[2] The test matrix is shown in Table 1. The helmets used were a standard lightweight HGU-55/P and a variable weighted impact helmet (VWI), which is an HGU-55/P modified for weight attachment at various center-of-gravity locations. In this study, the VWI helmet was used to simulate the mass properties of current Air Force front-loaded helmet-mounted systems. The seat back and seat pan of the sled were not reclined. A PCU-15/P or 16/P harness and HBU lap belt were used to restrain the subjects, with anchor points preloaded to 20 ± 5 lbs. A photograph of the HIA facility (also referred to as the "sled track") is shown in Figure 1. Prior to human testing, tests were conducted at each test condition with an instrumented manikin.

Table 1. Test Matrix

G Level	Helmet Weight (lb)					
	0.0 (no helmet)	**2.0**	**3.0**	**3.5**	**4.0**	**4.5**
6			T2		A2	C
7						W
8			L, LR*	V	B	D
10	E	X	F	M		

*Cell LR required the subjects to use a lighter brace, approximating 15-25 lbs of force on the headrest

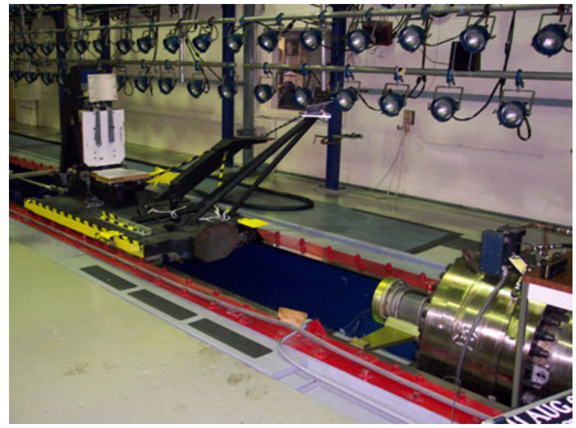

Figure 1. Horizontal Impulse Accelerator (HIA)

Measured electronic data included sled velocity and accelerations, seat accelerations, subject head and torso accelerations and displacements, and forces developed in the seat and the restraint system. The head accelerations were measured with a triaxial accelerometer array mounted on a bite bar. The bite bar accelerometer array also contained an angular accelerometer to record rotational acceleration about the y-axis. Neck loads and moments seen at the occipital condyles (OC, or head-neck joint) were calculated using the measured head accelerations, subject anthropometry, and helmet inertial properties. The measured headrest pre-impact loads were also statistically evaluated for these cells. The headrest pre-impact load is the magnitude of force applied to the headrest by the subject just prior to impact as a result of required bracing techniques. The subjects were instructed to brace at approximately 30 lbs of force against the headrest, although many subjects braced at somewhat higher levels. A real-time headrest load monitor was used so that subjects could adjust their brace accordingly. For cell LR, the subjects were instructed to brace at a lower level of 15-25 lbs of force.

RESULTS

A total of 269 human tests were evaluated following outlier removal. Of particular interest were the neck loads. In general, shear (X) neck

loads increased linearly with increasing sled (input) acceleration (Figure 2). It was observed that although subjects were instructed to brace at a constant level for all test cells except LR, the average male pre-impact bracing level increased with increasing sled acceleration (impact) level (Figure 3). This was not the case with the female subjects however, as a decrease in bracing level was observed from 8 to 10 G. Overall neck loads increased slightly with an increase in helmet weight, except at 4.0 lbs which showed a decrease in loads for both male and female subjects as compared to the 3.5 lb helmet (Figure 4). The pre-impact bracing averages across increasing helmet weights are shown in Figure 5.

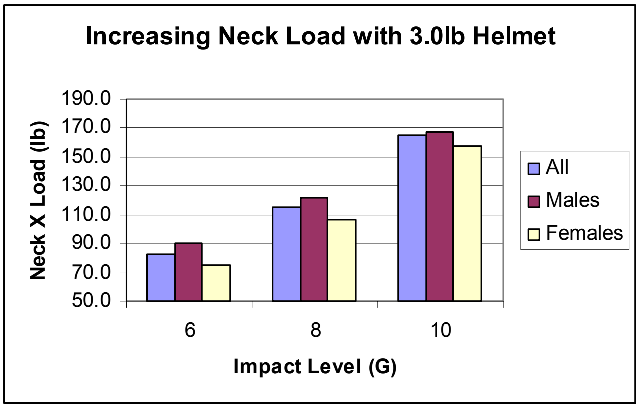

Figure 2. Neck X Load Increasing with Sled Impact Level

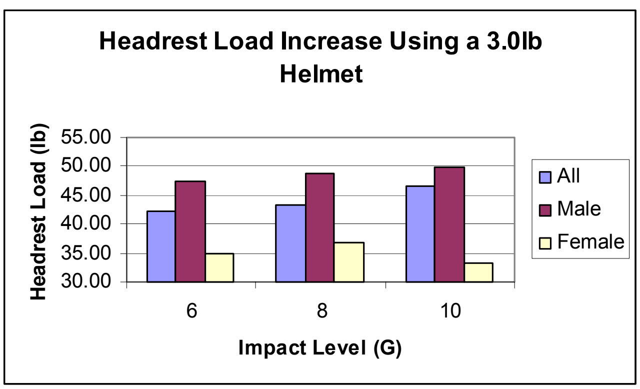

Figure 3. Pre-Impact Headrest Load Increasing with Sled Impact Level

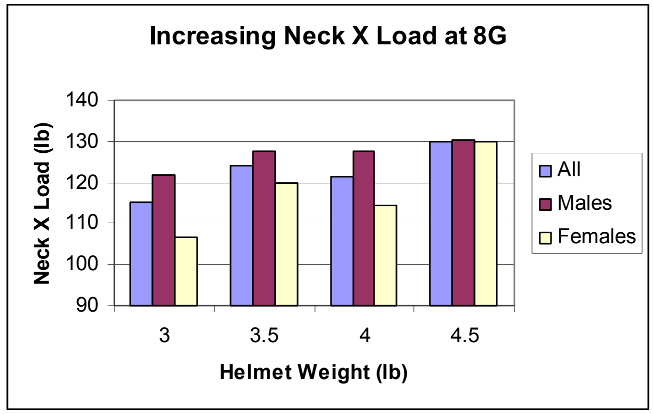

Figure 4. Neck X Load with Increasing Helmet Weight at 8G

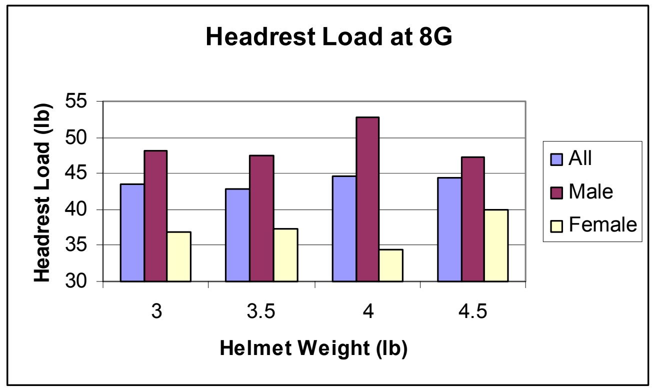

Figure 5. Pre-Impact Headrest Load with Increasing Helmet Weight at 8G

The neck loads calculated at the experimental levels were used to create a model to predict neck loads for seat accelerations and helmet weights outside the ranges used in this study. To do this, neck load data from those subjects who had completed a pre-determined range of test parameters for both sled acceleration (6-8 G) and helmet weight (3-4.5 lbs) were included in the analysis. Twenty-six subjects met the criteria for evaluation. (Cell LR was omitted due to the use of different bracing levels.) Using the analysis of covariance model, each subject's neck X load was predicted to determine minimum, mean and maximum values across subjects at seat accelerations up to 15 G and helmet weights up to 5.5 lbs. Actual values where available were used in place of predicted values. As shown in Figure 6, the neck X load predictions for the 26 subjects at 15 Gx and 5.5

lbs helmet ranged from 273 to 350 lbs with a mean of 301 lbs. Each subject's predicted value has a 95% prediction interval that is approximately ± 45 lbs. Variability around the predictions increases this range conservatively to values from 228 to 395 lbs. This model assumed a linear relationship between neck X load and both input acceleration level and helmet weight. The predictive equation for means in Figure 6 is:

$$\text{Mean X Force} = -22.257 + 14.075*Gx - 3.743*Wgt + 1.612*Gx*Wgt$$

Where 'Gx' is the input acceleration and 'Wgt' is total weight of the helmet.

Rearranging this equation we can factor out Gx:

$$14.075*Gx + 1.612*Gx*Wgt = 1.612*Gx*(Wgt+8.731)$$

Therefore, X Force = -22.257 + 1.612*Gx*(Wgt+8.731) - 3.743*Wgt

In physical terms, the 8.731 constant corresponds to the weight of the head and the 1.612 multiplication factor is due to the fact that the head acceleration overshoots the impact acceleration. Previous test programs have shown that the percentage of overshoot can vary with the pulse duration.

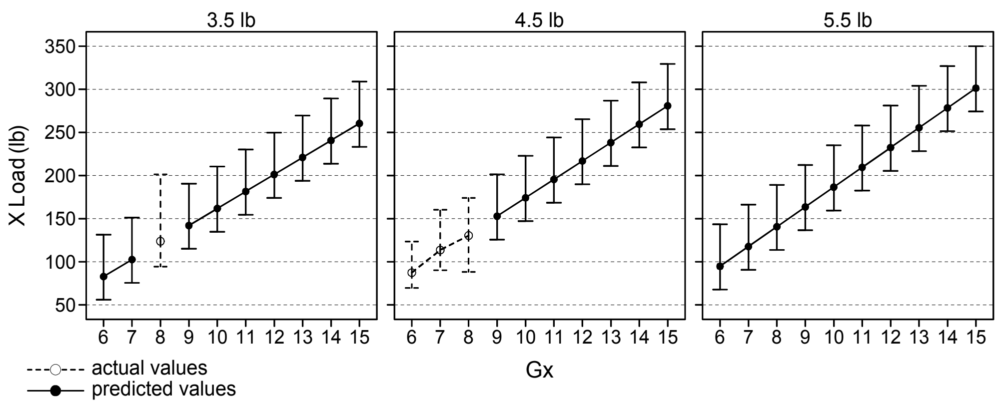

Figure 6. Neck X Load versus Sled Acceleration Level. Whiskers represent the minimum and maximum values across the 26 subjects.

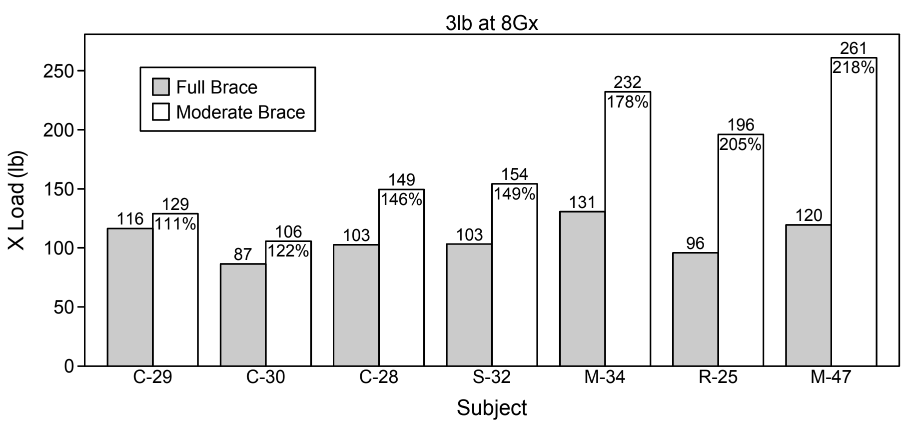

Figure 7. Bracing Effects on Neck X Load at 8G with 3.0 lb Helmet

Effects of subject bracing: The neck loads from cell LR, which required the subjects to brace moderately before impact, were compared to the loads of cell L, which called for a full brace. All subjects experienced higher neck loads in cell LR, two of whom had neck loading that more than doubled what they experienced during a full brace with the same testing conditions. On average, subjects experienced 52% higher neck loads with a moderate brace as compared to a full brace with the same impact and helmet parameters (Figure 7). The average pre-impact headrest loads (bracing level) at cells L and LR were 43 lbs and 26 lbs, respectively. The bracing reduced the maximum head acceleration which resulted in a lower neck load.

DISCUSSION

Neck loads: Neck stiffness or soreness was reported in approximately 15% of the tests, mostly at higher acceleration levels with heavier helmets. Females were more likely than males to experience these symptoms. Three male subjects completed cell M (10 G, 3.5-lb helmet), all of whom experienced neck pain, and had an average neck X load of 167 lbs. The helmet weight was subsequently limited to 3.0 lb during 10 G runs. We recommend using these values as the maximum limits for acceleration input and helmet weight for experimental frontal impact sled testing. The maximum neck X load observed in the male subject population was 265 lbs, occurring in cell F (10 G, 3.0 lb helmet). The maximum neck X load in the female subject population was 190 lbs and occurred at cell X (10 G, 2.0 lb helmet). These were extreme values and not typical in all of the subjects.

As expected, neck loads increased linearly with increasing impact acceleration. This linearity was extrapolated out to include higher acceleration levels and heavier helmets. Using this method the predicted X forces at 15 G and 5.5 lb helmet were within the range of 273-350 lbs. However, the possibility exists for individuals to experience non-linear loading trends at the higher impact accelerations, in which case the neck loads could be even larger than estimated here.

The neck loads from this study were compared to known injury limits set by previous cadaver studies conducted in the same axis. In 1971 Mertz and Patrick conducted a study that involved a human volunteer and four cadaver specimens on an impact sled track.[3] The human subject was accelerated up to 9.6 G with approximately 5 lbs of total helmet weight. A maximum shear load of 177.5 lbs was experienced by the subject; and neck pain was noted which produced stiffness that lasted several days. The cadavers were accelerated up to 14.2 G with the added helmet weight. A maximum shear load of 473.0 lbs was observed in the cadaver specimens without any observable ligamentous, disc, or bone damage as noted from x-ray analysis of the neck structures. These cadaver neck loads were higher than the predicted values of neck loads for our human subjects under comparable conditions of 15 G seat acceleration and 5.5 lbs helmet weight. The difference is probably due to our subjects being able to brace their heads against the headrest, thus decreasing their maximum neck force. The Mertz and Patrick study also employed a headrest that was extended several inches forward which likely contributed to greater head rotation and acceleration. Our human data would therefore appear to be a more reliable approximation of actual pilot shear neck loads that would be experienced during the ejection sequence, particularly during extreme levels of the parachute opening shock phase.

There is concern about injury risk due to neck loads that fall between the Mertz and Patrick cadaver injury threshold of 473 lbs and our experimental maximum tolerable levels of 190 lbs (females) and 265 lbs (males), due to the incidence of pain reported by some of our

subjects when experiencing neck loads at or below these levels. Based on our subjects' symptoms and depending on subject anthropometry and gender, it is conceivable that severe pilot whiplash-type injuries could occur within this range of neck loads. Although it is unlikely that permanently debilitating neck injury would be incurred at these levels, these injuries could nonetheless cause incapacitation of the pilot over an extended period.

Headrest bracing: Pre-impact headrest loads increased with increasing impact acceleration. The subjects were informed of the impending level before each test; therefore, motivation could have been the cause for the increase in bracing level. As shown in Figure 7, the bracing level had an obvious effect on neck loads. Since only one female participated in the moderate bracing cell, results from this observation are based on a primarily male population. In reviewing the remaining cells requiring a full brace, the female subjects braced at lower levels on average than the males, but did not necessarily experience higher neck loading. The bracing reduced the maximum head acceleration which resulted in a lower neck load. The head motion was reduced relative to the seat, and the head acceleration began to approximate the impact acceleration.

In general, there was a notable increase in the incidence of sore necks within the moderate bracing cell as compared to the full bracing cell conducted under the same conditions. An interesting finding was that some of the male subjects experienced significant soreness when bracing at 20-25 lbs force against the headrest, while many of the female subjects who were bracing at this level during the full brace cells did not experience any soreness. Some of the subjects also had difficulty adequately holding their braces during the impact when asked to brace at less than their maximum level. It appears that maximum bracing acts to offset neck loading during impact and is more effective in preventing neck soreness than moderate bracing. This appears to be true regardless of the level of the subject's maximum bracing. We therefore recommend that pilots perform a maximum brace during the catapult phase of ejection as well as just prior to the opening of their parachute.

CONCLUSION

The data from this study provide insight into the mechanisms and thresholds of human neck response during frontal impact. Since a pain threshold was reached with the few subjects who tested at 10 G with a 3.5 lb helmet, we recommend 10 G sled acceleration and 3.0 lb helmet weight as the maximum limits for experimental frontal impact sled testing. Our extrapolated data indicate that even under extreme conditions as seen during the parachute opening phase of aircraft ejection, pilots would likely not incur significant permanent neck injury. However, there is concern that severe whiplash could occur under certain adverse conditions, particularly with helmet weights greater than 5.0 lbs, that could result in temporary incapacitation of the pilot. The likelihood of injury would depend on the pilot's gender, anthropometry, and bracing techniques. Future work is planned that will more thoroughly investigate these injury risk factors.

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REFERENCES

1. Buhrman, J.R., C.E. Perry, and F.S. Knox III (1994). Human and Manikin Head/Neck Response to +Gz Acceleration When Encumbered by Helmets of Various Weights. *Aviation, Space, and Environmental Medicine*, *65*, 1086-1090.

2. Buhrman, J. R., C.E. Perry, and S.E. Mosher (2000). A Comparison of Male and Female

Acceleration Responses During Laboratory Frontal –Gx Axis Impact Tests. *AFRL-HE-WP-TR-2001-0022*.

3. Mertz, H.J. and L.M. Patrick (1971). Strength and response of the human neck. *Proceedings of the Fifteenth Stapp Car Crash Conference*, Society of Automotive Engineers, Warrendale, PA: SAE.

4. Perry, C.E. (1998). The Effect of Helmet Inertial Properties on Male and Female Head Response During +Gz Impact Accelerations. *SAFE Journal, 28(1), 32-38.*

5. Perry, C.E. and J.R. Buhrman (1997). Head Mounted Display (HMD) Head and Neck Biomechanics. In J.E. Melzer and K. Moffitt (Eds.), *Head Mounted Displays: Designing for the User (Chap.6, pp 147-174).* New York: McGraw-Hill.

6. Perry, C.E. and J.R. Buhrman (1996). Effect of Helmet Inertial Properties of the Biodynamics of the Head and Neck During +Gz Impact Accelerations. *SAFE Journal*, 26(2), 34-41.

7. Perry, C.E., J.R. Buhrman, E.J. Doczy and S.E. Mosher (2003). Evaluation of the Effects of Variable Helmet Weight on Human Response During Lateral +Gy Impact. *AFRL-HE-WP-TR-2004-0013.*

8. Perry, C.E., A.R. Rizer, J.S. Smith, and B. Anderson (1997). Biodynamic Modeling of Human Neck Response During Vertical Impact. *SAFE Journal, 27(3), 183-191.*

9. Pint, S. M. (2003). Evaluation of the Human Response to Upper Torso Retraction with Added Helmet Weight. Abstract presented at AsMA Annual Scientific Meeting, San Antonio TX.

BIOGRAPHIES

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